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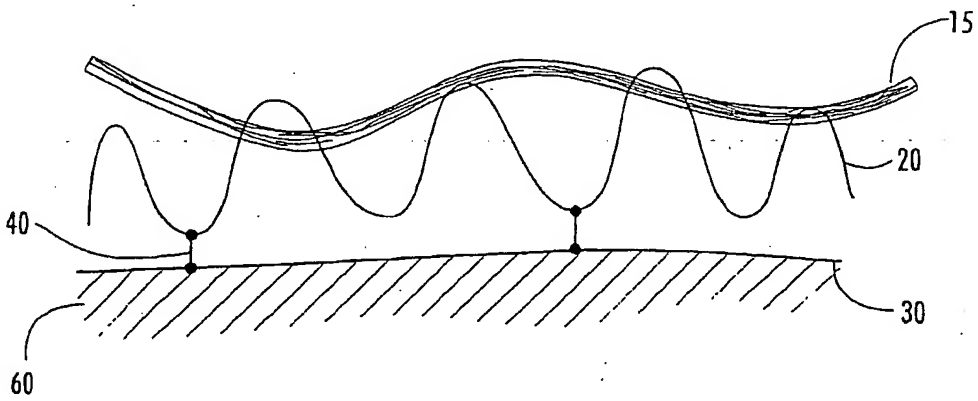

(57) Abstract: The present invention is directed to a stabilized bioactive hydrogel matrix coating for substrates, such as medical devices. The invention provides a coated substrate comprising a substrate having a surface, and a bioactive hydrogel matrix layer overlying the surface of the medical device, the hydrogel matrix comprising a first high molecular weight component and a second high molecular weight component, the first and second high molecular weight components each being selected from the group consisting of polyglycans and polypeptides, wherein at least one of the first and second high molecular weight components is immobilized (e.g., by covalent cross-linking) to the surface of the substrate.

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IMMOBILIZED BIOACTIVE HYDROGEL MATRICES AS SURFACE COATINGS

FIELD OF THE INVENTION

The present invention relates to cross-linked bioactive hydrogel matrices that are appropriate for use as immobilized bioactive coatings to improve the integration and performance of medical devices.

BACKGROUND OF THE INVENTION

The replacement of damaged or diseased tissues or organs by implantation has been, and continues to be, a long-standing goal of medicine towards which tremendous progress has been made. In addition, much progress has also been made in the field of treating patients with medical conditions through the implantation of therapeutic medical devices, such as glucose sensors and pacemakers. However, one of the most serious problems restricting the use of implants is the wound healing response elicited by implanted foreign materials (Ratner, B.D., "Reducing capsular thickness and enhancing angiogenesis around implant drug release systems" *Journal of Controlled Release* 78:211-218 (2002)).

Biocompatibility is defined as the appropriate response of the host to a foreign material used for its intended application. Biocompatibility further refers to the interaction between the foreign material and the tissues and physiological systems of the patient treated with the foreign material. Protein binding and subsequent denaturation as well as cell adhesion and activation have been invoked as determinants of a material's biocompatibility. Biocompatibility also implies that the implant avoids detrimental effects from the host's various protective systems and remains functional for a significant period of time. With respect to medical devices, biocompatibility is determined to a large extent by the type of acute reaction provoked by implantation. The extent to which a medical device is integrated with the surrounding tissue depends upon the type of wound healing response that is evoked by the implanted material. *In vitro* tests designed to assess cytotoxicity or protein

binding are routinely used for the measurement of a material's potential biocompatibility. In other words, the biocompatibility of a material is dependent upon its ability to be fully integrated with the surrounding tissue following implantation.

The modulation of this tissue response to an implanted medical device comprised of a foreign material is pivotal to successful implantation and performance of such medical devices. Mammalian systems recognize foreign materials, such as surgically implanted objects or medical devices. Upon binding to sites on these foreign materials, a cascade of events occur that notify inflammatory cells to surround such materials and initiate a series of wound healing events which ultimately lead to the formation of an avascular fibrous capsule surrounding the implanted device. The formation of an avascular fibrous capsule can severely limit the life and usefulness of the implanted medical device, especially in situations where direct contact with specific tissue, such as vascular tissue, muscle tissue, or nerve tissue is vital to the effectiveness of the device.

Previous research has shown that the specific interactions between cells and their surrounding extracellular matrix play an important role in the promotion and regulation of cellular repair and replacement processes (Hynes, S.O., "Integrins: a family of cell surface receptors" *Cell* 48:549-554 (1987)). Consequently, there has been a heightened interest in work related to biocompatible polymers useful in therapeutic applications. One particular class of polymers that have proven useful for such applications, including contact lens materials, artificial tendons, matrices for tissue engineering, and drug delivery systems, is hydrogels (Wheeler JC, Woods JA, Cox MJ, Cantrell RW, Watkins FH, Edlich RF.; Evolution of hydrogel polymers as contact lenses, surface coatings, dressings, and drug delivery systems.; J Long Term Eff Med Implants. 1996;6(3-4):207-17 and Schacht, E., "Hydrogels prepared by crosslinking of gelatin with dextran dialdehyde" *Reactive & Functional Polymers* 33:109-116 (1997)). Hydrogels are commonly accepted to be materials consisting of a permanent, three-dimensional network of hydrophilic polymers with water filling the space between the polymer chains, and they may be obtained by copolymerizing suitable hydrophilic monomers, by chain extension, and by cross-linking hydrophilic pre-polymers or polymers.

Prior work has shown that a thermoreversible hydrogel matrix, which is liquid near physiologic temperatures, elicits vasculogenesis and modulates wound healing in

dermal ulcers (Usala AL, Dudek R, Lacy S, Olson J, Penland S, Sutton J, Ziats NP, Hill RS: Induction of fetal-like wound repair mechanisms in vivo with a novel matrix scaffolding. Diabetes 50 (Supplement 2): A488 (2001); and Usala AL, Klann R, Bradfield J, Ray S, Hill RS, De La Sierra D, Usala M, Metzger M, Olson G: Rapid Induction of vasculogenesis and wound healing using a novel injectable connective tissue matrix. Diabetes 49 (Supplement 1): A395 (2000)). This bioactive hydrogel material has also been shown to improve the healing in response to implanted foreign materials; demonstrating a decrease in the surrounding fibrous capsule thickness and a persistent increase in blood supply immediately adjacent to implanted materials exposed to this thermoreversible hydrogel (Ravin AG, Olbrich KC, Levin LS, Usala AL, Klitzman B.; Long- and short-term effects of biological hydrogels on capsule microvascular density around implants in rats. J Biomed Mater Res. 2001 May 1;58(3):313-8.). However the use of such a bioactive thermoreversible hydrogel as a biomaterial coating for a medical device is not practical for devices requiring three-dimensional or thermal stability. Accordingly, there is a need for a bioactive material that is stable at body temperatures and thus appropriate for use as a coating for use with medical devices, particularly those intended for implantation into mammals.

BRIEF SUMMARY OF THE INVENTION

The invention provides a coated substrate, comprising a substrate having a surface, and a bioactive hydrogel matrix layer overlying the surface of the substrate and immobilized thereon, the hydrogel matrix layer comprising a first high molecular weight component and a second high molecular weight component, the first and second high molecular weight components each being selected from the group consisting of polyglycans and polypeptides. As used herein, the term "immobilized" refers to affixation of one or more components of the hydrogel matrix layer via any chemical or mechanical bonding force or process, such as by covalent attachment. The hydrogel matrix coating may further comprise one or more enhancing agents selected from the group consisting of polar amino acids, amino acid analogues, amino acid derivatives, intact collagen, and divalent cation chelators.

Preferred substrates include medical devices. The bioactive hydrogel compositions are useful both as a layer that serves as a structural component of a medical device and as a bioactive hydrogel coating that modulates the wound healing

response to an implanted device and improves tissue integration of a medical device. As a structural component of a medical device, bioactive hydrogel-coated biomaterials can be designed as a space-filling scaffold used to direct tissue organization and vascularization of a medical device. One exemplary use would be as a composite wound healing device comprising a polymeric microbial barrier and an immobilized bioactive hydrogel of sufficient thickness to provide a three dimensional structure to fill anatomic voids such as those resulting from donor-site tissue harvesting. As a functional coating of a medical device, bioactive hydrogel coatings are expected to reduce the avascular capsule surrounding an implanted device, and improve the intimate contact between surrounding tissues and active device elements and hence the performance of devices such as implanted glucose sensors for closed-loop control of diabetes. The compositions are also useful as bioactive hydrogel coatings for artificial organs containing functional tissue cells, and other passive or active medical devices or implants, and other biosensors.

Also provided is a method of preparing a coated substrate, such as a coated medical device. The method comprises immobilizing a first high molecular weight component on the surface of the substrate, wherein the first molecular weight component is selected from the group consisting of polyglycans and polypeptides. The first high molecular weight component is contacted with a second high molecular weight component also selected from the group consisting of polyglycans and polypeptides. The contacting step occurs before, during or after the immobilizing step. The two high molecular weight components form an immobilized bioactive hydrogel coating on the surface of the substrate. Preferably, one of the high molecular weight components is a polyglycan, such as dextran, and the other is a polypeptide, such as gelatin.

In a preferred embodiment, the immobilizing step comprises covalently attaching at least one of the high molecular weight components to the surface of the substrate. One or more of the high molecular weight components and/or the surface can be chemically modified, such as by oxidation or amination, to form reactive sites thereon capable of participating in covalent bonding. The high molecular weight components can be modified to comprise a plurality of pendant reactive groups along the backbone of the molecule or a single reactive group located at each terminus thereof.

BRIEF DESCRIPTION OF THE DRAWINGS

Having thus described the invention in general terms, reference will now be made to the accompanying drawings, which are not necessarily drawn to scale, and wherein:

Figure 1 illustrates formation of open alpha chains derived from collagen monomers;

Figure 2A illustrates the effect of the association of the alpha chains with dextran;

Figure 2B illustrates the behavior of the alpha chains without association of the dextran;

Figure 3 illustrates the effect of other hydrogel matrix additives;

Figure 4A illustrates a polyglycan immobilized to a surface of a medical device;

Figure 4B illustrates a polyglycan immobilized to a surface of a medical device and a polypeptide associated with the polyglycan to form a hydrogel;

Figure 5 illustrates graphically the effect of a hydrogel matrix in promoting cell aggregation;

Figure 6 illustrates graphically the effect of a hydrogel matrix through induction of transforming growth factor beta 3;

Figure 7 illustrates a polyglycan immobilized to a surface of a medical device through a terminal group and a polypeptide associated with the polyglycan to form a hydrogel;

Figure 8 illustrates one method of forming an immobilized bioactive hydrogel matrix of the present invention; and

Figure 9 illustrates a covalently cross-linked hydrogel matrix.

DETAILED DESCRIPTION OF THE INVENTION

The present invention now will be described more fully hereinafter with reference to the accompanying drawings, in which preferred embodiments of the invention are shown. This invention may, however, be embodied in many different forms and should not be construed as limited to the embodiments set forth herein; rather, these embodiments are provided so that this disclosure will be thorough and

complete, and will fully convey the scope of the invention to those skilled in the art. Like numbers refer to like elements throughout..

The formulation of a thermoreversible hydrogel matrix providing a cell culture medium and composition for preserving cell viability is taught by U.S. Patent No. 6,231,881, herein incorporated by reference in its entirety. Additionally, a hydrogel matrix useful in promoting vascularization is provided in U.S. Patent No. 6,261,587, herein incorporated by reference in its entirety. The thermoreversible hydrogel matrix taught by these references is a gel at storage temperatures and molten at physiologic temperatures, and comprises a combination of a collagen-derived component, such as gelatin, a long chain polyglycan, such as dextran, and effective amounts of other components, such as polar amino acids. The thermoreversible hydrogel matrix taught by these references is discussed below in connection with Figures 1-3.

Collagen is a major protein component of the extracellular matrix of animals. Collagen is assembled into a complex fibrillar organization. The fibrils are assembled into bundles that form the fibers. The fibrils are made of five microfibrils placed in a staggered arrangement. Each microfibril is a collection of collagen rods. Each collagen rod is a right-handed triple-helix, each strand being itself a left-handed helix. Collagen fibrils are strengthened by covalent intra- and intermolecular cross-links which make the tissues of mature animals insoluble in cold water. When suitable treatments are used, collagen rods are extracted and solubilized where they keep their conformation as triple-helices. This is denatured collagen and differs from the native form of collagen, but has not undergone sufficient thermal or chemical treatment to break the intramolecular stabilizing covalent bonds found in collagen. When collagen solutions are extensively heated, or when the native collagen containing tissues are subjected to chemical and thermal treatments, the hydrogen and covalent bonds that stabilize the collagen helices are broken, and the molecules adopt a disordered conformation. By breaking these hydrogen bonds, the polar amine and carboxylic acid groups are now available for binding to polar groups from other sources or themselves. This material is gelatin and is water-soluble at 40-45°C.

As noted above, gelatin is a form of denatured collagen, and is obtained by the partial hydrolysis of collagen derived from the skin, white connective tissue, or bones of animals. Gelatin may be derived from an acid-treated precursor or an alkali-treated precursor. Gelatin derived from an acid-treated precursor is known as Type A, and

gelatin derived from an alkali-treated precursor is known as Type B. The macromolecular structural changes associated with collagen degradation are basically the same for chemical and partial thermal hydrolysis. In the case of thermal and acid-catalyzed degradation, hydrolytic cleavage predominates within individual collagen chains. In alkaline hydrolysis, cleavage of inter-and intramolecular cross-links predominates.

Figure 1 illustrates the hydrolytic cleavage of the tropocollagen **10**, forming individual polar alpha chains of gelatin **15**. Heating tropocollagen **10** disrupts the hydrogen bonds that tightly contain the triple stranded monomers in mature collagen.

Figures 2A-2B illustrate stabilization of the matrix monomeric scaffolding by the introduction of a long-chain polyglycan, such as dextran **20**. As depicted in Figure 2A, the dextran **20** serves to hold open the gelatin **15**, that has been previously heated, by interfering with the natural predisposition of the gelatin **15** to fold upon itself and form hydrogen bonds between its polar groups. In the absence of dextran **20**, as shown in Figure 2B, when the gelatin **15** begins to cool, it will form hydrogen bonds between the amino and carboxylic acid groups within the linear portion of the monomer and fold upon itself, thus limiting available sites for cellular attachment.

The thermoreversible matrix contains a polyglycan, such as dextran, at a therapeutically effective concentration ranging from, for example, about 0.01 to about 10 mM, preferably about 0.01 to about 1 mM, most preferably about 0.01 to about 0.1 mM. In one embodiment, dextran is present at a concentration of about 0.09 mM.

The thermoreversible matrix also contains gelatin, at a therapeutically effective concentration ranging from, for example, about 0.01 to about 40 mM, preferably about 0.05 to about 30 mM, most preferably about 1 to 5 mM. Advantageously, the gelatin concentration is approximately 1.6 mM.

In order to increase cell binding, intact collagen may be added in small amounts to the thermoreversible matrix in order to provide additional structure for the cells contained in the matrix. The final concentration of intact collagen is from about 0 to about 5 mM, preferably about 0 to about 2 mM, most preferably about 0.05 to about 0.5 mM. In one embodiment, the concentration of intact collagen is about 0.11 mM.

The thermoreversible matrix may additionally contain an effective amount of polar amino acids, which are commonly defined to include tyrosine, cysteine, serine,

threonine, asparagine, glutamine, asparatic acid, glutamic acid, arginine, lysine, and histidine. For application in the present invention, the amino acids are preferably selected from the group consisting of cysteine, arginine, lysine, histidine, glutamic acid, aspartic acid and mixtures thereof, or derivatives or analogues thereof. By amino acid is intended all naturally occurring alpha amino acids in both their D and L stereoisomeric forms, and their analogues and derivatives. An analog is defined as a substitution of an atom or functional group in the amino acid with a different atom or functional group that usually has similar properties. A derivative is defined as an amino acid that has another molecule or atom attached to it. Derivatives would include, for example, acetylation of an amino group, amination of a carboxyl group, or oxidation of the sulfur residues of two cysteine molecules to form cystine. The total concentration of all polar amino acids is generally between about 3 to about 150 mM, preferably about 10 to about 65 mM, and more preferably about 15 to about 40 mM.

Advantageously, the added polar amino acids comprise L-cysteine, L-glutamic acid, L-lysine, and L-arginine. The final concentration of L-glutamic acid is generally about 2 to about 60 mM, preferably about 5 to about 40 mM, most preferably about 10 to about 20 mM. In one embodiment, the concentration of L-glutamic acid is about 15 mM. The final concentration of L-lysine is generally about 0.5 to about 30 mM, preferably about 1 to about 15 mM, most preferably about 1 to about 10 mM. In one embodiment, the concentration of L-lysine is about 5.0 mM. The final concentration of L-arginine is generally about 1 to about 40 mM, preferably about 1 to about 30 mM, most preferably about 5 to about 15 mM. In one embodiment, the final concentration of arginine is about 10 mM. The final concentration of L-cysteine, which provides disulfide linkages, is generally about 5 to about 500 μM, preferably about 10 to about 100 μM, most preferably about 15 to about 25 μM. In one embodiment, the final concentration of cysteine is about 20 μM.

The thermoreversible matrix is preferably based upon a physiologically compatible buffer, one embodiment being Medium 199, a common nutrient solution used for *in vitro* culture of various mammalian cell types (available commercially from Sigma Chemical Company, St. Louis, MO), which is further supplemented with additives and additional amounts of some medium components, such as supplemental amounts of polar amino acids as described above.

Advantageously, aminoguanidine may be added to this formulation; however, other L-arginine analogues may also be used in the present invention, such as N-monomethyl L-arginine, N-nitro-L-arginine, or D-arginine. The final concentration of aminoguanidine is generally about 5 to about 500 μM, preferably about 10 to about 100 μM, most preferably about 15 to about 25 μM. In one embodiment, the final concentration is about 20 μM.

Additionally, the matrix may include one or more divalent cation chelators, which increase the rigidity of the matrix by forming coordinated complexes with any divalent metal ions present. The formation of such complexes leads to the increased rigidity of the matrix by removing the inhibition of hydrogen bonding between $-NH_2$ and –COOH caused by the presence of the divalent metal ions. A preferred example of a divalent cation chelator that is useful in the present invention is ethylenediaminetetraacetic acid (EDTA) or a salt thereof. The concentration range for the divalent cation chelator, such as EDTA, is generally about 0.01 to about 10 mM, preferably 1 to about 8 mM, most preferably about 2 to about 6 mM. In a one embodiment, EDTA is present at a concentration of about 4 mM.

Figure 3 illustrates the effect of polar amino acids and L-cysteine added to stabilize the units **25**, formed by the gelatin **15** and dextran **20**, by linking the exposed monomer polar sites to, for example, arginine's amine groups or glutamic acid's carboxylic acid groups. Furthermore, disulfide linkages can be formed between L-cysteine molecules (thereby forming cystine), which in turn form hydrogen bonds to the gelatin **15**.

The mechanical and thermal characteristics of the thermoreversible hydrogel described above are to a large extent determined by the thermomechanical properties of one of its major components, gelatin. Gelatin-based matrices typically are molten at near physiologic temperatures and hence cannot be expected to have the requisite durability and mechanical properties when required for implantation as a medical device in certain applications. Therefore, it is imperative to stabilize these gels through a variety of intermolecular interactions including hydrogen bonding, electrostatic or polar amino acid mediated bonding, hydrophobic bonding and covalent bonding. Although not wishing to be bound by theory, it is believed that the types of bonding mechanisms described above in association with a polyglycan stabilize polypeptides such as gelatin. For example, as discussed in more detail

below, the positively charged polar groups of the collagen-derived alpha chains are then able to associate with the negatively charged hydroxyl groups of the repeating glucose units found in, for example, dextran. The gelatin and dextran form a composite bioactive hydrogel containing macromolecular proteoglycan-type structures.

Unlike the prior art thermoreversible matrix discussed above, the present invention provides stabilized compositions comprising an immobilized bioactive matrix that can be used, for example, as a coating for implanted medical devices to modulate localized wound healing around an implanted medical device, or to produce a localized vasculogenic response and encourage tissue integration with the implanted device. The present invention is also directed to a method for manufacturing an immobilized bioactive coating or film of cell scaffolding material directly on a substrate surface, such as the surface of a medical device. The present invention provides a cell attachment scaffold that supports the initiation of a series of cell signaling pathways and modulates the localized wound healing and acute inflammatory cascade in response to the implanted foreign material. By "bioactive" is intended the ability to facilitate or discourage a cellular or tissue response of a host to implanted materials. Examples include, but are not limited to, induction of vasculogenesis, inhibition of the formation of a foreign body response, controlled tissue reorganization around an implanted material or medical device, promotion of cellular adhesion, or regeneration of specific anatomic features such as dermal pegs and rete ridges during dermal healing. The term "stabilized" or "stable" is intended to refer to compositions that are water-swellable, poorly soluble, solid or semi-solid materials at physiological temperature (i.e., about 37°C) and in physiological fluids (e.g., aqueous body fluids having a physiological pH of about 7.4), which remain present in the host for sufficient time to achieve the intended response.

It is not believed that the immobilized or cross-linked bioactive matrix coating affects the intrinsic material or chemical properties of the underlying substrate (e.g., a medical device). Unlike prior art devices or hydrogels, the present invention is believed to modulate the acute response of a host animal to polymeric materials typically used for medical device manufacture, not by changing the material's properties, but rather by changing the localized tissue response to the implanted material.

The bioactive coatings of the invention can be applied to a surface of any substrate where such coatings would be useful. In particular, suitable substrates include medical devices. By medical device is intended to include any device, whether active or passive in nature, which may be inserted or implanted into a host organism, such as a mammal. The term "medical device" is further intended to encompass any natural or synthetic device or material, including nucleic acids, which is used therapeutically either *in vivo*, such as by implantation into a human or animal, or *ex vivo* to provide therapeutic benefit, whether intended to be a permanent implant or temporary implant. Such devices include but are not limited to catheters, artificial arteries, artificial organs, medical devices containing cells of either engineered tissues or isolated tissue fragments or cells derived from naturally occurring or genetically engineered sources, ligament replacements, bone replacements, glucose sensors, coronary pacemakers, lap-bands, monitors, artificial larynxes, prostheses (such as testicular, esophageal, tracheal, and fallopian tube), brain stimulators, bladder pacemakers, bladder stimulators, shunts, stents, tubes, defibrillators, cardioverters, heart valves, joint replacements, fixation devices, ocular implants, cochlear implants, breast implants, neurostimulators, bone growth stimulators, vascular grafts, muscle stimulators, left ventricular assist devices, pressure sensors, vagus nerve stimulators, drug delivery systems, sutures, staples, cell scaffolding materials, active or passive medical devices comprised of gels, pastes or solids and the like and *ex vivo* bioreactors for liver, kidney or other organ support devices. Ex vivo bioreactors are external to the patient's body and used temporarily to provide metabolic function pending organ transplantation or other therapeutic intervention. Any foreign object that is placed in the body, or in contact with body tissues or fluids whether for a temporary time period or permanently, may benefit from the present invention.

The medical device of the present invention may be rigid or flexible, solid, fibrillar, or woven and may be derived from naturally occurring materials or constructed from synthetic materials. Exemplary materials of construction include acrylates, polyglycolic-polylactic acid copolymers, polyhydroxybutyrates, polyesters (such as Dacron®), expanded polytetrafluoroethylene (ePTFE), bioactive glass, ceramics (such as hydroxyapatites), coralline materials, processed tissue (such as demineralized bone), polycarbonate, polyurethane/polycarbonate copolymers, metals (such as titanium), and mixtures, composites or subassemblies thereof. Bioactive

glasses generally contain silicon dioxide (SiO_2) as a network former and are characterized by their ability to firmly attach to living tissue. Examples of bioactive glasses available commercially and their manufacturers include Bioglass® (American Biomaterials Corp., USA, 45% silica, 24% calcium oxide (CaO), 24.5% disodium oxide (Na_2O), and 6% pyrophosphate (P_2O_5)), Consil® (Xeipon Ltd., UK), NovaBone® (American Biomaterials Corp.), Biogran® (Orthovita, USA), PerioGlass® (Block Drug Co., USA), and Ceravital® (E.Pfeil & H. Bromer, Germany). Corglaes® (Giltech Ltd., Ayr, UK) represents another family of bioactive glasses containing pyrophosphate rather than silicon dioxide as a network former. These glasses contain 42-49 mole% of P_2O_5, the remainder as 10-40 mole% as CaO and Na_2O.

The term "subassemblies" is intended to encompass multiple piece construction of the device, wherein the individual pieces of the device are constructed of the same or different materials. The term "composite" is intended to encompass devices comprising different active or passive materials, present to meet specific design requirements for the intended medical device.

The present invention provides a stabilized bioactive matrix layer or coating that overlies an exposed surface of a medical device or other substrate and is immobilized thereon. As the present invention is useful as a coating for any portion of a medical device that may have contact with body tissues or fluids, either *in vivo* or *ex vivo*, both temporarily and permanently, the term "exposed surface" is intended to encompass any such surface of a medical device that is exposed to brief or prolonged contact with body tissues or fluids. The word "surface" as used throughout in reference to a medical device or other substrate is therefore intended to encompass, in particular, any surface of the medical device operatively positioned for exposure to body tissues or fluids.

The matrix layer is formed from at least two high molecular weight components. The high molecular weight components of the bioactive hydrogel matrix are selected from the group consisting of high molecular weight polyglycans, high molecular weight polypeptides, and combinations thereof. By high molecular weight polyglycan is intended any polysaccharide consisting of more than about 10 monosaccharide residues joined to each other by glycosidic linkages. The polyglycan may consist of the same monosaccharide residues, or various monosaccharide residues or derivatives of monosaccharide residues. Dextran, a preferred

polysaccharide, typically comprises linear chains of α(1→6)-linked D-glucose residues, often with α(1→2)- or α(1→3)-branches. Native dextran, produced by a number of species of bacteria of the family Lactobacilliaceae, is a polydisperse mixture of components.

The polyglycan component preferably has a molecular weight range of about 2,000 to about 8,000,000 Da, more preferably about 20,000 to about 1,000,000 Da. Unless otherwise noted, molecular weight is expressed herein as number average molecular weight (M_n), which is defined as $\frac{\sum NiMi}{\sum Ni}$, wherein Ni is the number of polymer molecules (or the number of moles of those molecules) having molecular weight Mi.

Any polysaccharide, including glycosaminoglycans (GAGs) or glucosaminoglycans, with suitable viscosity, molecular mass and other desirable properties may be utilized in the present invention. By glycosaminoglycan is intended any glycan (i.e., polysaccharide) comprising an unbranched polysaccharide chain with a repeating disaccharide unit, one of which is always an amino sugar. These compounds as a class carry a high negative charge, are strongly hydrophilic, and are commonly called mucopolysaccharides. This group of polysaccharides includes heparin, heparan sulfate, chondroitin sulfate, dermatan sulfate, keratan sulfate, and hyaluronic acid. These GAGs are predominantly found on cell surfaces and in the extracellular matrix. By glucosaminoglycan is intended any glycan (i.e. polysaccharide) containing predominantly monosaccharide derivatives in which an alcoholic hydroxyl group has been replaced by an amino group or other functional group such as sulfate or phosphate. An example of a glucosaminoglycan is poly-N-acetyl glucosaminoglycan, commonly referred to as chitosan. Exemplary polysaccharides that may be useful in the present invention include dextran, heparan, heparin, hyaluronic acid, alginate, agarose, carageenan, amylopectin, amylose, glycogen, starch, cellulose, chitin, chitosan and various sulfated polysaccharides such as heparan sulfate, chondroitin sulfate, dextran sulfate, dermatan sulfate, or keratan sulfate.

By high molecular weight polypeptide is intended any tissue-derived or synthetically produced polypeptide, such as collagens or collagen-derived gelatins. Although collagen-derived gelatin is the preferred high molecular weight polypeptide

component, other gelatin-like components characterized by a backbone comprised of sequences of amino acids having polar groups that are capable of interacting with other molecules can be used. For example, keratin, decorin, aggrecan, glycoproteins (including proteoglycans), and the like could be used to produce the polypeptide component. In one embodiment, the polypeptide component is porcine gelatin from partially hydrolyzed collagen derived from skin tissue. Polypeptides derived from other types of tissue could also be used. Examples include, but are not limited to, tissue extracts from arteries, vocal chords, pleura, trachea, bronchi, pulmonary alveolar septa, ligaments, auricular cartilage or abdominal fascia; the reticular network of the liver; the basement membrane of the kidney; or the neurilemma, arachnoid, dura mater or pia mater of the nervous system. Purified polypeptides including, but not limited to, laminin, nidogen, fibulin, and fibrillin or protein mixtures such as those described by U.S. Patent No. 6,264,992 and U.S. Patent No. 4,829,000, extracts from cell culture broth as described by U.S. Patent No. 6,284,284, submucosal tissues such as those described in U.S. Patent No. 6,264,992, or gene products such as described by U.S. Patent No. 6,303,765 may also be used. Another example of a suitable high molecular weight polypeptide is a fusion protein formed by genetically engineering a known reactive species onto a protein. The polypeptide component preferably has a molecular weight range of about 3,000 to about 3,000,000 Da, more preferably about 30,000 to about 300,000 Da.

In a preferred embodiment, gelatin and dextran are components of the bioactive matrix of the present invention. For ease of describing the invention, the terms "gelatin" and "dextran" are used throughout with the understanding that various alternatives as described above, such as other polyglycan and polypeptide components readily envisioned by those skilled in the art, are contemplated by the present invention.

Figure 4A illustrates one embodiment of the present invention wherein a high molecular weight component of the matrix **20**, such as a polysaccharide (e.g., dextran), is immobilized to an exposed surface **30** of a medical device **60**. In this embodiment, the high molecular weight component **20** is attached to the exposed surface **30** via a plurality of covalent linkages **40**, such as peptide linkages, between the exposed surface **30** of the medical device **60** and pendant reactive groups along the high molecular weight component chain **20**. In this manner, a high molecular

weight component of the matrix, such as either dextran or gelatin, can be covalently attached to an exposed surface **30** of a medical device **60** to form an immobilized coating.

In this particular embodiment, the surface of the medical device must first be activated. Surface activation of synthetic materials is well known to those skilled in the art of surface modification. For example, surface activation methods are commonly used for the immobilization of biomacromolecules during the preparation of affinity chromatography media. Common surface modification techniques are outlined in *Affinity Chromatography: A Practical Approach*, Dean *et al.*, IRL Press, 1985 ISBN 0-904147-71-1, which is incorporated by reference in its entirety. Other methods of preparing synthetic or naturally derived surfaces for subsequent reaction with macromolecular species in solution are well-known to those skilled in the art.

In one embodiment, reactive amine groups are formed on the surface. For example, perfluorinated poly(ethylene-*co*-propylene) tape (Teflon®) and poly(ethylene terephthalate) (PET) sheets can be coated with thin amine polymer layers deposited from a "monomer" vapor of volatile amines using a radio-frequency glow discharge. The density of the formed amine layer can be varied by selecting appropriate volatile amines. In one particular study, low amine density films were prepared using *n*-heptylamine, while high amine density films were prepared using allylamine (See, Kingshott *et al.*, "Effects of cloud-point grafting, chain length, and density of PEG layers on competitive adsorption of ocular proteins" *Biomaterials* 23:2043-2056 (2002)). The carboxyl groups of activated dextran or gelatin react with the available amine groups of the surface, to form a Schiff base which can then be further reduced using either sodium borohydride or sodium cyanoborohydride to form peptide links. The high molecular weight component is thus immobilized on the surface of the medical device by covalent linkages therebetween.

The extent and uniformity of surface coverage by the immobilized high molecular weight components can be varied using reaction parameters well known to those skilled in the art. Similarly, by varying the concentration of reactive species in solution above an activated surface, the thickness of the immobilized bioactive hydrogel may be controlled. For example, a thin uniform bioactive hydrogel coating may be desirable for the improved long-term function of an implanted glucose biosensor, where the intended device function requires rapid equilibration between the

local tissue environment and the sensor interface for optimal performance. Such methods are well known to those skilled in the art. In another example, relatively thick bioactive hydrogel coatings may be desirable for medical devices requiring extensive tissue integration for optimal performance. Cell scaffolds or tissue bulking devices for sphincter repair and regeneration are examples of such medical devices which may benefit from a design composed of an underlying substrate coated with a bioactive hydrogel coating to provide both structural and mechanical tissue support while encouraging tissue integration and localized tissue regeneration. Using methods outlined above, one can construct bioactive hydrogel coatings ranging in thickness from about 10^{-4} to about 10 cm.

The immobilized dextran or gelatin component may be used as a template upon which a cell scaffolding material similar to the thermoreversible hydrogel matrix described above may be constructed. For example, at least one additional high molecular weight component (e.g., gelatin), and at least one enhancing agent may be added to the immobilized high molecular weight component (e.g., dextran) to form an immobilized bioactive hydrogel matrix on the surface of the medical device. The relative amounts of the various hydrogel ingredients may be varied to obtain a wide range of desirable therapeutic and biomechanical properties. In one embodiment, the same concentrations as used in the thermoreversible matrix formulation discussed above are used.

By "enhancing agent" or "stabilizing agent" is intended any compound added to the hydrogel matrix, in addition to the two high molecular weight components, that enhances the hydrogel matrix by providing further stability or functional advantages. Suitable enhancing agents, which are admixed with the high molecular weight components and dispersed within the hydrogel matrix, include many of the additives described earlier in connection with the thermoreversible matrix discussed above. The enhancing agent can include any compound, especially polar compounds, that, when incorporated into the cross-linked hydrogel matrix, enhances the hydrogel matrix by providing further stability or functional advantages.

Preferred enhancing agents for use with the stabilized cross-linked hydrogel matrix include polar amino acids, amino acid analogues, amino acid derivatives, intact collagen, and divalent cation chelators, such as EDTA or salts thereof. Polar amino acids is intended to include tyrosine, cysteine, serine, threonine, asparagine,

glutamine, aspartic acid, glutamic acid, arginine, lysine, and histidine. The preferred polar amino acids are L-cysteine, L-glutamic acid, L-lysine, and L-arginine. Suitable concentrations of each particular enhancing agent are the same as noted above in connection with the thermoreversible hydrogel matrix. Polar amino acids, EDTA, and mixtures thereof, are preferred enhancing agents. The enhancing agents can be added to the matrix composition before, during, or after immobilization of a high molecular weight component to the surface of the medical device.

The enhancing agents are particularly important in the stabilized cross-linked bioactive hydrogel matrix because of the inherent properties they promote within the matrix. The hydrogel matrix exhibits an intrinsic bioactivity that will become more evident through the additional embodiments described hereinafter. It is believed the intrinsic bioactivity is a function of the unique stereochemistry of the cross-linked macromolecules in the presence of the enhancing and strengthening polar amino acids, as well as other enhancing agents.

For example, aggregation of human fibroblasts exposed to bioactive hydrogels has been observed, while aggregation is not observed when fibroblasts are exposed to the individual components of the bioactive hydrogel. Results from numerous (over fifty) controlled experiments have shown that normal neonatal human skin fibroblasts form multi-cell aggregates when exposed to the complete thermoreversible hydrogel formulation at 37 °C, while no such cell aggregating activity is demonstrated using formulations in which the bioactive copolymer is not formed. The aggregated cells form tightly apposed cell clusters with interdigitating cytoplasmic processes, while cells treated with formulations lacking the copolymer remain round and without surface projections. As shown in Figure 5, in a sample of human fibroblasts exposed to a bioactive hydrogel comprising dextran and gelatin, at least 80% of the cells present were in an aggregated state while less than 20% of the cells present remained as single cells. The opposite effect was observed in samples where the human fibroblasts were exposed to collagen monomer alone, carbohydrate alone, or were left untreated. In samples exposed to collagen monomer alone, approximately 75% of the cells remained in a single cell configuration while only about 25% of the cells were in an aggregated state. Nearly the same effect was observed in samples exposed to carbohydrate alone. In samples that were left untreated, approximately 60% of the

cells remained in a single cell state while only about 40% of the cells were in an aggregated state.

In a preferred embodiment, dextran is immobilized on an exposed surface of the device and gelatin is added, thereby forming a copolymer with the dextran through hydrogen bonding and polar interactions. This embodiment is shown in Figure 4B where dextran **20**, having gelatin **15** associated therewith, is immobilized on an exposed surface **30** of a medical device **60**. These interactions may then be further stabilized through subsequent covalent bonding mediated by added reactive species (i.e., enhancing agents). The finished product is a stabilized, bioactive hydrogel which functions as a cell attachment scaffold having a localized effect on cellular responses, thereby improving the long-term performance of the medical device.

One such effect on cellular response is illustrated in Figure 6, which provides a graphical representation of the results of one study of gene expression in normal human neonatal skin fibroblasts. That study demonstrated a marked induction of transforming growth factor beta 3 (TGF-β3) following hydrogel exposure. Expression of this gene is associated with scarless wound healing as seen during fetal development. Conversely, in the same cells, transforming growth factor beta 1 (TGF-β1), which is instrumental in scar formation during adult wound healing, was not induced by hydrogel exposure reflecting the ability of the hydrogel to facilitate a specific character of response in a population of tissue cells.

In this embodiment, where dextran is immobilized on the surface and gelatin is added, the dextran, containing predominantly relatively unreactive hydroxyl groups, requires activation to convert the hydroxyl groups to the more reactive aldehyde groups suitable for cross-linking to the surface. This must be done prior to contacting the surface of the medical device, which has previously undergone surface modification, such as by the method described above for forming reactive amine groups. For instance, the dextran, or other polyglycan component, can be modified, such as by oxidation, in order to cross-link with the modified surface of the medical device. One known reaction for oxidizing polysaccharides is periodate oxidation. The basic reaction process utilizing periodate chemistry is well known and appreciated by those skilled in the art. Periodate oxidation is described generally in *Affinity Chromatography: A Practical Approach*, Dean, *et al.*, IRL Press, 1985 ISBN0-904147-71-1. The oxidation of dextran by the use of periodate-based

chemistry is described in U.S. Patent Nos. 3,947,352 and 6,011,008, which are herein incorporated by reference in their entirety.

In periodate oxidation, hydrophilic matrices may be activated by the oxidation of the vicinal diol groups. With a cellulosic surface, or other polysaccharide surface, this is generally accomplished through treatment with an aqueous solution of a salt of periodic acid, such as sodium periodate ($NaIO_4$), which oxidizes the sugar diols to generate reactive aldehyde groups (e.g. dialdehyde residues). This method is a rapid, convenient alternative to other known oxidation methods, such as those using cyanogen bromide. Materials activated by periodate oxidation may be stored at 4 °C for several days without appreciable loss of activity. This method can be used to prepare activated biomaterial surfaces appropriate for polypeptide binding or to prepare soluble activated polysaccharides to be bound to surfaces containing primary amine groups.

Periodate oxidized materials, such as dextran, readily react with materials containing amino groups, such as an activated surface of a medical device or a polypeptide, producing a cross-linked material through the formation of Schiff's base links. A Schiff base is a name commonly used to refer to the imine formed by the reaction of a primary amine with an aldehyde or ketone. The aldehyde groups formed on the cellulosic surface react with most primary amines between pH values from about 4 to about 6. The Schiff's base links form between the dialdehyde residues on the cellulosic surface and the free amino groups on the polypeptide or activated surface of the medical device. The cross-linked product may subsequently be stabilized (i.e. formation of stable amine linkages) by reduction with a borohydride, such as sodium borohydride ($NaBH_4$) or cyanoborohydride ($NaBH_3CN$). The residual aldehyde groups may be consumed with ethanolamine. Other methods known to those skilled in the art may be utilized to provide reactive groups on one of the high molecular weight components of the matrix.

The immobilized hydrogel matrix coatings of the present invention are biomimetic, meaning the coating layer imitates or stimulates a biological process or product. Some biomimetic processes have been in use for several years, such as the artificial synthesis of vitamins and antibiotics. More recently, additional biomimetic applications have been proposed, including nanorobot antibodies that seek and destroy disease-causing bacteria, artificial organs, artificial arms, legs, hands, and

feet, and various electronic devices. The biomimetic scaffolding materials of the present invention may yield therapeutically useful surface coatings that are stable at about 37 °C, or body temperature.

Once a high molecular weight component, such as dextran or gelatin, has been covalently cross-linked to the surface of the medical device, the second high molecular weight component can be added. The two high molecular weight components, one being covalently cross-linked to the surface of the medical device, interact through hydrogen bonding and polar attractions, thereby forming a stabilized copolymer. Additionally, at least one enhancing agent, as described above, can be added to further stabilize the hydrogel matrix.

Dextran or gelatin may be immobilized in a pendant, or chain-like, fashion to the surface of the medical device as shown in Figures 4A and 4B. This approach may be useful for the development of glucose sensors or other in-dwelling devices requiring improved soft tissue integration for long-term function and biocompatibility. This configuration may also be useful for guided tissue growth or as a means of modulating cell growth and structure within a three-dimensional tissue engineered construct such as a device intended to function as an artificial liver or kidney. The use of a bio-erodible bulk material allows one to fabricate engineered constructs either for controlled drug delivery with long-term release of a pharmaceutical agent to an induced vascular bed, or the development of guided tissue growth for bulking applications.

In another embodiment, the dextran **20** (or gelatin component) may be attached to the surface via a peptide link at one terminus of the dextran chain, as shown in Figure 7. As with the embodiment shown in Figure 4, the surface **30** of the device **60** must be activated using common surface modification methods as outlined above. One skilled in the art would readily understand the parameters necessary for immobilizing dextran at one terminus. (See for example, Larm, O. *et al.*, "A New Non-Thrombogenic Surface Prepared By Selective Covalent Binding Of Heparin Via A Modified Reducing Terminal Residue" *Biomat Med Dev Artif Organs* 11:161-73 (1983)). One skilled in the art would also readily understand the methods for immobilizing other macromolecules such as polypeptides via one terminus. (See for example, Gregorius, K. *et al.*, "In Situ Deprotection: A Method For Covalent Immobilization Of Peptides With Well-Defined Orientation For Use In Solid Phase

Immunoassays Such As Enzyme-Linked Immunosorbent Assay" *Anal Biochem* 299:94-91 (2001), and Olbrich K.C. *et al.*, "Surfaces Modified With Covalently-Immobilized Adhesive Peptides Affect Fibroblast Population Motility" *Biomaterials* 17:144–153 (1996)). Those skilled in the art will further recognize that surface activation of substrates can achieve the same end of immobilizing one of the two high molecular weight components upon a surface without requiring modification of the native macromolecule. For example, surface immobilization of proteins to insoluble PVA substrates has been described previously (See, Manecke G. and Vogt, H.G., *J Solid Phase Biochem* 4(233) (1979)).

Dextran immobilization occurring through activated terminal groups forms "end-on" immobilized dextran, or "brush-like surfaces" as shown in Figure 7. Here, the gelatin monomer **15** may then be formed around the immobilized dextran **20** to produce a biomimetic structure **80** designed to facilitate the activation of cell signaling pathways similar to those found during embryonic development. This configuration may lead to a more hydrogel-like surface and may provide a "softer" surface for guided cell growth. As with the pendant configuration, both permanent and bio-erodible surfaces may be modified in this manner. The extent of surface coverage by dextran is dependent on the molecular weight of the dextran and the extent of dextran branching.

Figure 8 illustrates yet another embodiment, whereby gelatin is immobilized to the surface of a medical device. Gelatin with its native reactive primary amines distributed along the polypeptide backbone can be immobilized to surfaces containing aldehyde groups. Activated surfaces may be formed using radio frequency glow discharge treatment of polymeric surfaces in the presence of oxygen or other reactive oxidative species. This surface treatment forms aldehydes and other reactive species on the surface of the treated material, which can subsequently react with and immobilize gelatin directly. Gelatin may also form linkages, such as peptide linkages, with the surface of the device in a pendant fashion or only at a terminus of the gelatin chain. Gelatin is immobilized or tethered to a surface of a medical device at room temperature as shown in Step 1 of Figure 8. Next, dextran is added and the temperature elevated to alter the gelatin quaternary structure to disrupt thermally stabilized intramolecular mediated hydrogen bonding producing a more open polypeptide conformation as shown in Step 2. For Step 2, the surface temperature

should be elevated to at least about 30 to about 90°C, preferably about 40 to about 60°C. Various reactive species including, but not limited to, polar amino acids and amino acid derivatives as described earlier are also added and allowed to react at this step. Decreasing the temperature may further assist intermolecular interactions so that the dextran begins interacting with the gelatin as shown in Step 3. During Step 3, additional components, such as polar amino acids, can also be added to the matrix. Finally, in Step 4, the surface is once again at room temperature and a stabilized bioactive matrix coating is formed on the surface. The resulting matrix is a tightly bonded gelatin/dextran bioactive hydrogel matrix. Subsequent post-processing steps may include washing to remove excessive reactive species from the stabilized hydrogel. Typically, the temperature of the surface of the medical device varies from about 20 to about 60 °C during the above-described steps. The pH range for the copolymer formation, as described above, is within the physiologic range, preferably between about 6 and about 8, most preferably between about 7 and about 7.6.

Another method useful for immobilizing macromolecules, such as gelatin or dextran, to a surface is by way of a mechanical process. For example, a synthetic thermoplastic polymer may be partially swollen in the presence of a fluid containing a polymer solvent dispersed in water. The addition of macromolecules to this fluid allows the added solutes to become entrapped within the open, swollen surface of the polymer. By rapidly exchanging the fluid phase surrounding the swollen polymer, the polymer de-swells, entrapping the added macromolecular solutes within the surface of the polymer.

In yet another embodiment, bioactive hydrogels may be formed directly using electrooxidation. In this method, a molten thermoreversible hydrogel is placed in an electrolytic cell containing two conductive electrodes. A potential difference is applied between the electrodes, and oxidizable species in solution (i.e. functional groups such as hydroxyls, and amines) are directly oxidized at the anode. The resulting reactive oxidized compounds condense at the anodic surface to form a water-insoluble hydrogel coating. In this manner, for example, titanium mesh commonly used for craniofacial reconstructive surgery can be coated with an immobilized bioactive hydrogel to direct tissue organization and vascularization at the site of the implant.

Additional methods for immobilizing macromolecules are known in the art and include the methods described in the following references, each of which is incorporated by reference in its entirety: (i) Puleo D.A. *et al.*, "A technique to immobilize bioactive proteins, including bone morphogenetic protein-4 (BMP-4), on titanium alloy" *Biomaterials*, 23:2079-2087 (2002) ; (ii) Kong, U. *et al.*, "Durable Surface Modification of Poly(tetrafluoroethylene) by Low Pressure H_2O Plasma Treatment Followed by Acrylic Acid Graft Polymerization" *Coll Surf B: Biointerface* 24:63-71 (2002); (iii) Chandy, T. *et al.*, "Use of Plasma Glow for Surface Engineering Biomolecules to Enhance Blood Compatibility of Dacron and PTFE Vascular Prostheses" *Biomaterials* 21:699-712 (2000); (iv) Bos, G.W. *et al.*, "Proliferation of Endothelial Cells on Surface-Immobilized Albumin-Heparin Conjugate Loaded with Basic Fibroblast Growth Factor" *J Biomed Mater Res* 44:330-340 (1999); (v) Ayhan F. *et al.*, "Optimization of Urerase[sic] Immobilization onto Non-Porous HEMA Incorporated Poly(EGDMA) Microbeads and estimation of kinetic parameters" *Biores Technol* 81:131-40 (2002); (vi) Massia S.P. *et al.*, "Surface Immobilized Dextran Limits Cell Adhesion and Spreading" *Biomaterials* 21:2253-2261 (2000); (vii) Barie, N. *et al.*, "Covalent Photo-Linker Mediated Immobilization Of An Intermediate Dextran Layer To Polymer-Coated Surfaces For Biosensing Applications" *Bios Bioelect* 13:855-860 (1998); (viii) Chevolot, Y., *et al.*, "Immobilization On Polysytrene Of Diazirine Derivatives Of Mono- And Disaccharides: Biological Activities Of Modified Surfaces" *Bioorganic & Med Chem* 9:2943-53 (2001); (ix) Tsai, C.C. *et al.*, "Effects Of Heparin Immobilization On The Surface Characteristics Of A Biological Tissue Fixed With A Naturally Occurring Crosslinking Agent (Genipin) An In Vitro Study" *Biomaterials* 22:523-33 (2001); (x) Ito, Y., "Micropattern Immobilization Of Polysaccharide" *J Bioinorg Chem* 79:88-81 (2000); (xi) Massia, S.P. *et al.*, "Immobilized rgd Peptides On Surface-Grafted Dextran Promote Biospecific Cell Attachment" *J Biomed Mater Res* 56:390–399 (2001); and (xii) Dai L., *et al.*, "Biomedical Coatings By Covalent Immobilization Of Polysaccharides Onto Gas-Plasma-Activated Polymer Surfaces" *Surf Interface Anal* 29:46–55 (2000).

In yet another embodiment of the present invention, the two high molecular weight components of the hydrogel matrix surface coating may be cross-linked. As when cross-linking the dextran, or another polyglycan, to the modified surface of the

medical device, the dextran must also first be modified in order to cross-link with the gelatin component. For example, partial oxidation of dextran using sodium meta-periodate produces a polyaldehyde dextran that can be immobilized on amine derivatized surfaces. Dextran immobilization in the presence of sodium cyanoborohydride catalytically reduces the formed Schiff base to the more stable amide covalent bond. The immobilized dextran coated substrate can then be washed to remove the excess reagents, and treated with sodium meta-periodate to form additional aldehydes. This tethered polyaldehyde dextran can then cross-link with another high molecular weight component, such as gelatin.

The presence of cross-linking between the two high molecular weight components of the hydrogel coating is illustrated in Figure 9. As shown, in addition to being covalently attached to the exposed surface of the medical device, the dextran **20** can be covalently crosslinked to gelatin **15** by linkages **70**, thereby forming a crosslinked network **50**. The linkages **70** either result from reaction of functional groups on the gelatin **15** with functional groups on the dextran **20**, or result from reaction of a bifunctional crosslinker molecule with both the dextran **20** and gelatin **15**. One method of crosslinking gelatin and dextran is to modify the dextran molecules **20**, such as by oxidation, in order to form functional groups suitable for covalent attachment to the gelatin **15**. Dextran is modified, such as by oxidation, and stabilized via covalent bonding to gelatin **15**, thereby forming a cross-linked network **50**.

As noted above, periodate oxidation is one example of a known reaction for oxidizing polysaccharides that can also be used in this embodiment of the present invention in addition to other embodiments described previously. The reaction scheme can be carried out as before, oxidizing the sugar diols of the polyglycan, thereby forming reactive aldehyde groups. In this embodiment, the Schiff's base links form between the reactive aldehyde groups and the free amino groups on the polypeptide component of the hydrogel matrix. The cross-linked product may then be stabilized (i.e. formation of stable amine linkages) by reduction with a borohydride, such as sodium borohydride ($NaBH_4$) or cyanoborohydride ($NaBH_3CN$), and the residual aldehyde groups may be consumed with ethanolamine.

As an alternate method for forming the cross-linked hydrogel coating, a multifunctional cross-linking agent may be utilized as a reactive moiety that

covalently links the gelatin and dextran chains. Such bifunctional cross-linking agents may include glutaraldehyde, epoxides (e.g. bis-oxiranes), oxidized dextran, p-azidobenzoyl hydrazide, N-[α-maleimidoacetoxy]succinimide ester, p-azidophenyl glyoxal monohydrate, Bis-[ß-(4-azidosalicylamido)ethyl]disulfide (BASED), bis[sulfosuccinimidyl]suberate, dithiobis[succinimidyl propionate, disuccinimidyl suberate, 1-ethyl-3-[3-dimethylaminopropyl]carbodiimide hydrochloride, ethoxylated (20) trimethylpropane triacrylate, and other bifunctional cross-linking reagents known to those skilled in the art.

In one embodiment, 1.5 mL of a 0.5 mg/mL solution of Bis-[ß-(4-azidosalicylamido)ethyl]disulfide (BASED) in dimethyl sulfoxide (DMSO), is added to a foil-wrapped vessel containing 15 mL of liquid thermoreversible hydrogel as described above. Photoactivated non-specific cross-linking of the thermoreversible hydrogel occurs upon exposure of the reactive mixture to long-wavelength light, such as that provided by continuous exposure to a 550 watt bulb (flood light used in photography). Longer exposure times demonstrated better cross-linking.

In another embodiment utilizing a cross-linking agent, polyacrylated materials, such as ethoxylated (20) trimethylpropane triacrylate, may be used as a non-specific photo-activated cross-linking agent. Components of an exemplary reaction mixture would include thermoreversible hydrogel held at 39 °C, polyacrylate monomers, such as ethoxylated (20) trimethylpropane triacrylate, a photo-initiator, such as eosin Y, catalytic agents, such as 1-vinyl-2-pyrrolidinone, and triethanolamine. Continuous exposure of this reactive mixture to long-wavelength light (> 498 nm) would produce a cross-linked hydrogel network.

In a further embodiment of the present invention, both high molecular weight components are cross-linked to the surface of the medical device. In this embodiment, the surface of the medical device must be activated prior to contacting the high molecular weight components. For example, bis-oxiranes, such as 1,4-butanediol diglycidoxy ether react readily with hydroxy- or amino-containing biomaterials at alkaline pH to yield derivatives which possess a long-chain hydrophilic, reactive oxirane (epoxide), which, in turn, can be reacted with amines, hydroxyls and other nucleophiles. Oxirane-coupled ligands are widely used and extremely stable and the use of a long chain bis-oxirane reagent introduces a long hydrophilic spacer molecule between the

immobilized hydrogel components and the biomaterial surface which may be desirable in certain applications.

In another embodiment, the high molecular weight components may be modified in order to form reactive groups capable of reacting with the reactive groups of the activated surface of the medical device prior to contacting the surface of the medical device. This embodiment is not restricted by the order in which the high molecular weight components are contacted with the surface of the medical device. In one preferred embodiment, the surface of the medical device is activated, such as by radio frequency glow discharge in the presence of amine containing vapors to form reactive amine groups thereon, and modified dextran, such as oxidized dextran, is added. The dextran is covalently cross-linked to the surface of the medical device and gelatin is added. The gelatin is then also covalently cross-linked to the surface of the medical device through the polyaledehyde groups of the tethered dextran.

In another embodiment of the present invention, the surface of the medical device and the two high molecular weight components are all cross-linked to each other, wherein the polyglycan is covalently cross-linked to the surface of the medical device, the polypeptide is covalently cross-linked to the surface of the medical device, and the polyglycan and the polypeptide are cross-linked to each other. Various methods for carrying out this embodiment for coating a medical device would be envisioned by one skilled in the art. One possible method would comprise coating a polymeric medical device through radiation or electron beam grafting (See, Muzykewicz K.J. *et al.*, "Platelet adhesion and contact activation time tests on HEMA coated cellulose acetate membranes" *J Biomed Mater Res.* 9(5):487-99 (1975) and Venkataraman S. *et al.*, "The reactivity of alpha-chymotrypsin immobilized on radiation-grafted hydrogel surfaces" *J Biomed Mater Res.* 11(1):111-23 (1977)).

The stabilized cross-linked bioactive hydrogel can be used to encourage site-specific tissue regeneration, including vasculogenesis, in the area surrounding an implanted medical device with the stabilized cross-linked bioactive hydrogel immobilized thereon. It is known in the art to use intact collagen, gelatin, or dextran as a carrier to hold and deliver growth factors and the like in methods designed to promote tissue growth. (See, for example, Kawai, K. *et al.*, "Accelerated tissue Regeneration Through Incorporation of Basic Fibroblast Growth Factor-Impregnated Gelatin Microspheres into Artificial Dermis" *Biomaterials* 21:489-499 (2000); and

Wissink, M.J.B. *et al.*, "Binding and Release of Basic Fibroblast Growth Factor from Heparinized Collagen Matrices" *Biomaterials* 22:2291-2299 (2001)). By contrast, the intrinsic activity of the stabilized cross-linked hydrogel of the present invention is sufficient to elicit a specific sequence of biological responses, such as promoting tissue regeneration and vasculogenesis, without the addition of exogenous drugs or growth factors. In fact, the bioactive hydrogel matrix of the present invention can be substantially free, even completely free, of exogenous drugs or growth factors when used for vascularization or tissue regeneration. This intrinsically bioactive hydrogel, as a result of its unique structure, provides a cell attachment scaffold that modulates subsequent cellular activity, such as tissue regeneration and vasculogenesis.

The stabilized cross-linked hydrogel behaves similarly when used in other aspects of tissue regeneration. The hydrogel provides a stabilized structural lattice that facilitates cell retention and multiplication in areas with tissue damage. This is due in part to the intrinsic bioactivity of the hydrogel, which furthers the regenerative process. This is especially useful in applications where the success or functioning of an implanted medical device is dependent upon its integration with the surrounding tissue. The intrinsic bioactivity of the cross-linked hydrogel immobilized to the surface of the medical device not only reduces incidence of rejection by the host resulting from inflammatory response, immune response, etc., but it also increases healing and tissue regeneration in the site surrounding the implanted device.

The immobilized bioactive hydrogel matrix surface coating utilized in each of the embodiments described herein may be comprised solely of the two high molecular weight components. Preferably, each of the embodiments described herein incorporates additional components such as the enhancing agents utilized in the preferred embodiments described above. Table 1 below lists preferred components present within the immobilized bioactive hydrogel matrix surface coatings of the present invention along with suitable concentrations as well as preferred concentrations for each component. Note that the concentrations listed in Table 1 for gelatin and dextran would also be suitable for alternative polyglycan and polypeptide components.

Table 1

Component	Concentration Range	Preferred Concentration
L-glutamic acid	2 to 60 mM	15 mM
L-lysine	0.5 to 30 mM	5 mM
Arginine	1 to 40 mM	10 mM
Gelatin	0.01 to 40 mM	2 mM
L-cysteine	5 to 500 μM	20 μM
EDTA	0.01 to 10 mM	4 mM
Dextran	0.01 to 10 mM	0.1 mM

As noted above, the present invention provides numerous benefits including eliciting vascularization at a localized site, modulating localized wound healing response, and providing suitable means of developing a retrievable cell implantation device for cell-based therapeutics. Additional benefits may include the following: reduced scarring associated with degradation of bioerodible suture materials; improvement in the performance and long-term function of extravascular sensors such as glucose sensors routinely used for insulin delivery systems; improvement in the rate of healing, durability, and mechanical properties around structural implants such as artificial joints and tendons; reduced pain and associated complications arising from post surgical adhesions especially during abdominal or spinal injury; and improved integration between natural tissues and implanted structures (i.e. teeth, porous hydroxyapatite or ceramic materials for bone repair).

EXPERIMENTAL

The present invention is more fully illustrated by the following examples, which are set forth to illustrate the present invention and are not to be construed as limiting thereof.

Example 1

A beaker with an internal volume of 50 mL was equipped with two copper electrodes at a 2.5 cm separation. The beaker was filled with an aqueous solution of liquid thermoreversible hydrogel containing dextran and gelatin. A potential

difference of 18 V was applied across the cell. A hydrogel complex consisting of the covalently cross-linked thermoreversible hydrogel formulation immediately formed on the surface of the anode, and the thickness of the film increased with increasing time. The sterile hydrogel was insoluble in water at 37 °C, was adherent to the underlying substrate and conformed to the surface of the anodic metal.

One skilled in the art would readily recognize the utility of this method for producing adherent hydrogel coatings on metallic substrates such as titanium meshes used for reconstructive surgery. Such bioactive hydrogel coatings are expected to improve the vascularity and osteointegration of the implant.

Example 2

Activated biomaterial surfaces suitable for having the hydrogel matrix cross-linked thereto can be prepared by copolymerization of monomers containing bifunctional groups, one of which is protected. For example, the monomer glycidyl methacrylate can be copolymerized using free radical initiation with other acrylates to form hydrogels, and hydrogel films. Poly(2-hydroxyethyl methacrylate-co-glycidyl methacrylate)—poly(HEMA–GMA) hydrogel films can be prepared by UV-initiated photopolymerization with α,α'-azoisobutyronitrile (AIBN) as an initiator, preferably under an inert atmosphere at 25 °C. The epoxide content of the hydrogel films can be varied by varying the relative ratio of HEMA to GMA. For example, films with a high density of epoxides can be prepared by mixing 0.2 mL of HEMA, 0.8 mL GMA, 1 mL isopropyl alcohol, 10 mg AIBN (as a polymerization initiator), and 3.0 mL of 0.1M phosphate buffer (pH=7.0). The resulting mixture is stirred and equilibrated at 25 °C for 15 min in a thermostated water bath. The mixture can be then poured into the mold and exposed to long-wave ultraviolet radiation for 20 min. After polymerization, poly(HEMA–GMA) films can be washed several times with distilled water and cut into circular pieces with a biopsy punch. The functional epoxy group carrying poly(HEMA–GMA) film disks (10g wet weight, diameter=1.0 cm) formed as described above are equilibrated in phosphate buffer (50 mM, pH=8.0) for 2 hours, and transferred to a container holding the thermoreversible hydrogel held at 39 °C. Immobilization of the thermoreversible hydrogel to the surface of the biomaterial film can be carried out at 39 °C with frequent agitation. The poly(HEMA–GMA) films

coated with a thermoreversible hydrogel can be removed and washed to remove non-covalently attached hydrogel materials.

Many modifications and other embodiments of the invention will come to mind to one skilled in the art to which this invention pertains having the benefit of the teachings presented in the foregoing descriptions and the associated drawings. Therefore, it is to be understood that the invention is not to be limited to the specific embodiments disclosed herein and that modifications and other embodiments are intended to be included within the scope of the appended claims. Although specific terms are employed herein, they are used in a generic and descriptive sense only and not for purposes of limitation.

THAT WHICH IS CLAIMED:

1. A coated substrate, comprising:

a substrate having a surface; and

a bioactive hydrogel matrix layer overlying the surface of the substrate and immobilized thereon, the hydrogel matrix layer comprising a first high molecular weight component and a second high molecular weight component, the first and second high molecular weight components each being selected from the group consisting of polyglycans and polypeptides.

2. The coated substrate of Claim 1, wherein the substrate is a medical device.

3. The coated substrate of Claim 2, wherein the medical device is selected from the group consisting of active medical devices and passive medical devices.

4. The coated substrate of Claim 2, wherein the medical device is selected from the group consisting of ex vivo bioreactors for liver, kidney or other organ support systems, catheters, artificial arteries, artificial organs, tissue fragment-containing devices, cell-containing devices, ligament replacements, bone replacements, glucose sensors, coronary pacemakers, lap-bands, monitors, artificial larynxes, prostheses, brain stimulators, bladder pacemakers, shunts, stents, tubes, defibrillators, cardioverters, heart valves, joint replacements, fixation devices, ocular implants, cochlear implants, breast implants, neurostimulators, bone growth stimulators, vascular grafts, muscle stimulators, left ventricular assist devices, pressure sensors, vagus nerve stimulators, drug delivery systems, sutures, staples, and cell scaffolding materials.

5. The coated substrate of Claim 1, wherein the surface of the substrate is constructed of a material selected from the group consisting of acrylates, polyglycolic-polylactic acid copolymers, polyhydroxybutyrates, polyesters, expanded polytetrafluoroethylene (ePTFE), bioactive glass, ceramics, coralline materials,

processed tissue, polycarbonate, polyurethane/polycarbonate copolymers, metals, and mixtures, composites or subassemblies thereof.

6. The coated substrate of Claim 1, wherein at least one of the first and second high molecular weight components is covalently cross-linked to the exposed surface of the medical device.

7. The coated substrate of Claim 1, wherein the first high molecular weight component is a polyglycan and the second high molecular weight component is a polypeptide.

8. The coated substrate of Claim 7, wherein the polyglycan is a polysaccharide or a sulfated polysaccharide.

9. The coated substrate of Claim 8, wherein the polyglycan is a polysaccharide comprising more than about 10 monosaccharide residues joined to each other by glycosidic linkages.

10. The coated substrate of Claim 8, wherein the polysaccharide is selected from the group consisting of glycosaminoglycans and glucosaminoglycans.

11. The coated substrate of Claim 8, wherein the polysaccharide is selected from the group consisting of dextran, heparan, heparin, hyaluronic acid, alginate, agarose, carageenan, amylopectin, amylose, glycogen, starch, cellulose, and chitin.

12. The coated substrate of Claim 8, wherein the sulfated polysaccharide is selected from the group consisting of heparan sulfate, chondroitin sulfate, dextran sulfate, dermatan sulfate, and keratan sulfate.

13. The coated substrate of Claim 7, wherein the polyglycan has a molecular weight range of about 2,000 to about 8,000,000 Da.

14. The coated substrate of Claim 7, wherein the polyglycan has a molecular weight range of about 20,000 to about 1,000,000 Da.

15. The coated substrate of Claim 7, wherein the polypeptide is a tissue-derived or synthetic polypeptide.

16. The coated substrate of Claim 15, wherein the polypeptide is a tissue-derived polypeptide selected from the group consisting of collagens, gelatins, keratin, decorin, aggrecan, and glycoproteins.

17. The coated substrate of Claim 15, wherein the polypeptide is derived from tissue selected from the group consisting of submucosal tissues, arteries, vocal chords, pleura, trachea, bronchi, pulmonary alveolar septa, ligaments, auricular cartilage, abdominal fascia, liver, kidney, neurilemma, arachnoid, dura mater, and pia mater.

18. The coated substrate of Claim 15, wherein the polypeptide is selected from the group consisting of laminin, nidogen, fibulin, and fibrillin.

19. The coated substrate of Claim 7, wherein the polypeptide has a molecular weight range of about 3,000 to about 3,000,000 Da.

20. The coated substrate of Claim 7, wherein the polypeptide has a molecular weight range of about 30,000 to about 300,000 Da.

21. The coated substrate of Claim 7, wherein the polyglycan is dextran and the polypeptide is gelatin.

22. The coated substrate of Claim 21, wherein the dextran is present at a concentration of about 0.01 to about 10 mM.

23. The coated substrate of Claim 21, wherein the gelatin is present at a concentration of about 0.01 to about 40 mM.

24. The coated substrate of Claim 1, wherein the hydrogel matrix further comprises at least one enhancing agent, wherein the enhancing agent is selected from the group consisting of polar amino acids, amino acid analogues, amino acid derivatives, intact collagen, and divalent cation chelators.

25. The coated substrate of Claim 24, wherein the at least one enhancing agent comprises at least one polar amino acid selected from the group consisting of tyrosine, cysteine, serine, threonine, asparagine, glutamine, aspartic acid, glutamic acid, arginine, lysine, histidine, and mixtures thereof.

26. The coated substrate of Claim 25, wherein the polar amino acids are present at a concentration of about 3 to about 150 mM.

27. The coated substrate of Claim 25, wherein the polar amino acids are selected from the group consisting of L-glutamic acid, L-lysine, L-arginine, L-cysteine, and mixtures thereof.

28. The coated substrate of Claim 27, wherein the L-glutamic acid is present at a concentration of about 2 to about 60 mM.

29. The coated substrate of Claim 27, wherein the L-lysine is present at a concentration of about 0.5 to about 30 mM.

30. The coated substrate of Claim 27, wherein the L-arginine is present at a concentration of about 1 to about 40 mM

31. The coated substrate of Claim 27, wherein the L-cysteine is present at a concentration of about 5 to about 500 μM.

32. The coated substrate of Claim 24, wherein the at least one enhancing agent comprises a divalent cation chelator.

33. The coated substrate of Claim 32, wherein the divalent cation chelator is ethylenediaminetetraacetic acid or a salt thereof.

34. The coated substrate of Claim 33, wherein the ethylenediaminetetraacetic acid is present at a concentration of about 0.01 to about 10 mM.

35. The coated substrate of Claim 1, wherein the first high molecular weight component and the second high molecular weight component are covalently cross-linked to each other.

36. The coated substrate of Claim 1, wherein the matrix layer has a thickness of about 10^{-4} cm to about 10 cm.

37. A method of preparing a coated substrate, comprising the steps of:

providing a first high molecular weight component selected from the group consisting of polyglycans and polypeptides;

providing a substrate having a surface;

immobilizing the first high molecular weight component on the surface of the substrate;

contacting the first high molecular weight component with a second high molecular weight component selected from the group consisting of polyglycans and polypeptides, said contacting step occurring either before, during or after said immobilizing step, thereby forming an immobilized bioactive hydrogel coating on the surface of the substrate..

38. The method of Claim 37, further comprising, prior to said immobilizing step, chemically modifying the first high molecular weight component to form reactive sites thereon capable of participating in covalent bonding.

39. The method of Claim 38, wherein said modifying step comprises oxidizing the first high molecular weight component.

40. The method of Claim 39, wherein said modifying step comprises treating the first high molecular weight component with a salt of periodic acid.

41. The method of Claim 37, further comprising modifying the surface of the substrate to form reactive sites thereon capable of participating in covalent bonding prior to said immobilizing step.

42. The method of Claim 41, wherein modifying the surface of the substrate comprises forming a plurality of reactive amine groups on the surface.

43. The method of Claim 37, wherein the substrate is a medical device.

44. The method of Claim 43, wherein the medical device is selected from the group consisting of active medical devices and passive medical devices.

45. The method of Claim 43, wherein the medical device is selected from the group consisting of ex vivo bioreactors for liver, kidney or other organ support systems, catheters, artificial arteries, artificial organs, tissue fragment-containing devices, cell-containing devices, ligament replacements, bone replacements, glucose sensors, coronary pacemakers, lap-bands, monitors, artificial larynxes, prostheses, brain stimulators, bladder pacemakers, shunts, stents, tubes, defibrillators, cardioverters, heart valves, joint replacements, fixation devices, ocular implants, cochlear implants, breast implants, neurostimulators, bone growth stimulators, vascular grafts, muscle stimulators, left ventricular assist devices, pressure sensors, vagus nerve stimulators, drug delivery systems, sutures, staples, and cell scaffolding materials.

46. The method of Claim 37, wherein the surface of the substrate is constructed of a material selected from the group consisting of acrylates, polyglycolic-polylactic acid copolymers, polyhydroxybutyrates, polyesters, expanded polytetrafluoroethylene (ePTFE), bioactive glass, ceramics, coralline materials, processed tissue, polycarbonate, polyurethane/polycarbonate copolymers, metals, and mixtures, composites or subassemblies thereof.

47. The method of Claim 37, wherein the first high molecular weight component is a polyglycan and the second high molecular weight component is a polypeptide.

48. The method of Claim 47, wherein the polyglycan is a polysaccharide or a sulfated polysaccharide.

49. The method of Claim 48, wherein the polysaccharide is selected from the group consisting of dextran, heparan, heparin, hyaluronic acid, alginate, agarose, carageenan, amylopectin, amylose, glycogen, starch, cellulose, and chitin.

50. The method of Claim 48, wherein the sulfated polysaccharide is selected from the group consisting of heparan sulfate, chondroitin sulfate, dextran sulfate, dermatan sulfate, and keratan sulfate.

51. The method of Claim 47, wherein the polyglycan has a molecular weight range of about 2,000 to about 8,000,000 Da.

52 The method of Claim 47, wherein the polypeptide is a tissue-derived or synthetic polypeptide.

53. The method of Claim 47, wherein the polypeptide is a tissue derived polypeptide selected from the group consisting of collagens, gelatins, keratin, decorin, aggrecan, and glycoproteins.

54. The method of Claim 47, wherein the polypeptide is derived from tissue selected from the group consisting of submucosal tissues, arteries, vocal chords, pleura, trachea, bronchi, pulmonary alveolar septa, ligaments, auricular cartilage, abdominal fascia, liver, kidney, neurilemma, arachnoid, dura mater, and pia mater.

55. The method of Claim 47, wherein the polypeptide has a molecular weight range of about 3,000 to about 3,000,000 Da.

56. The method of Claim 47, wherein the polyglycan is dextran and the polypeptide is gelatin.

57. The method of Claim 37, further comprising contacting the first high molecular weight component with at least one enhancing agent, said contacting step occurring either before, during or after said immobilizing step, the enhancing agent being selected from the group consisting of polar amino acids, amino acid analogues, amino acid derivatives, intact collagen, and divalent cation chelators.

58. The method of Claim 57, wherein the at least one enhancing agent comprises at least one polar amino acid selected from the group consisting of tyrosine, cysteine, serine, threonine, asparagine, glutamine, aspartic acid, glutamic acid, arginine, lysine, histidine, and mixtures thereof.

59. The method of Claim 37, wherein the first high molecular weight component comprises a plurality of pendant reactive groups.

60. The method of Claim 37, wherein the first high molecular weight component comprises a reactive group located at a terminus thereof.

61. The method of Claim 37, wherein said immobilizing step comprises reacting carboxyl groups of the first high molecular weight component with amine groups on the surface of the substrate.

62. The method of Claim 37, wherein said immobilizing step comprises covalently cross-linking the first high molecular weight component to the surface of the substrate.

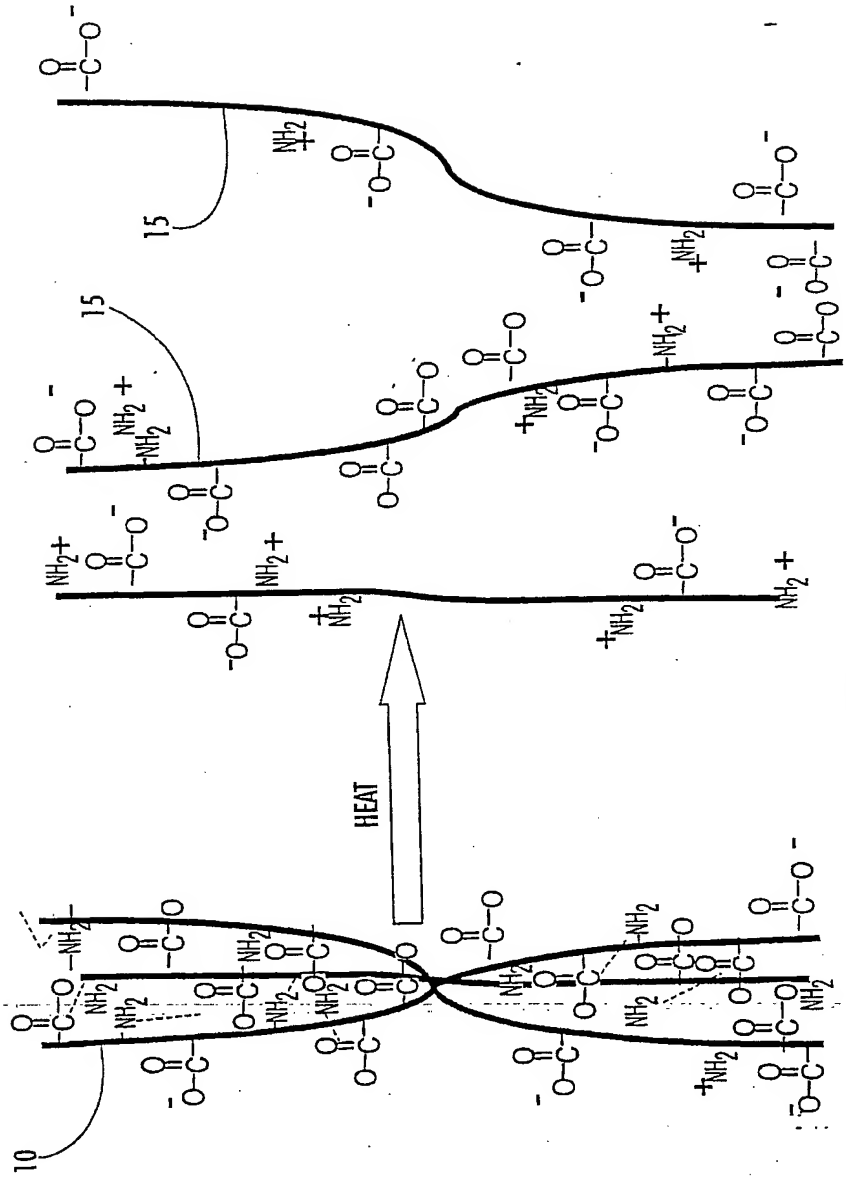


FIGURE 1

FIGURE 2A

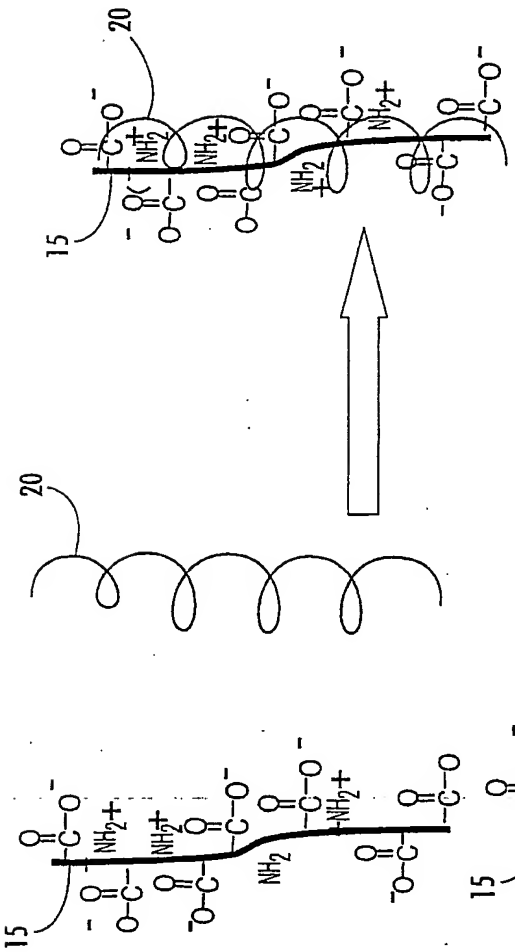
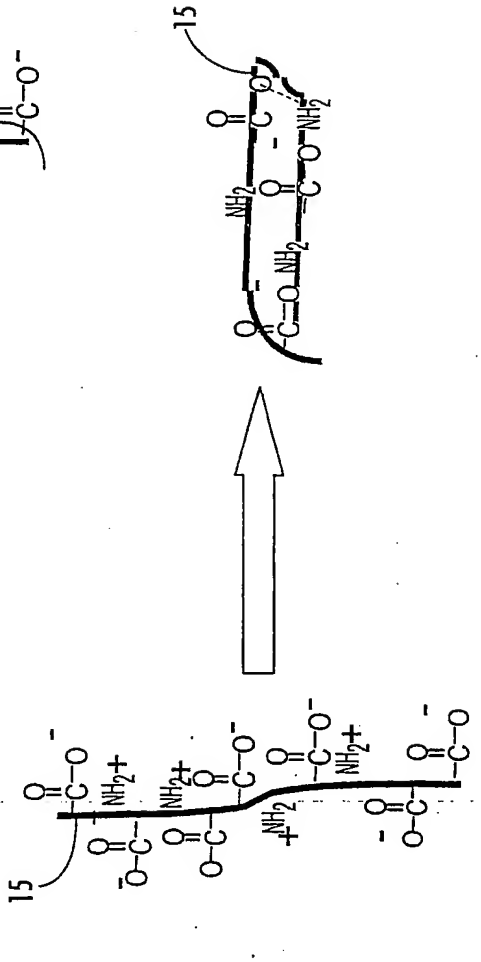


FIGURE 2B



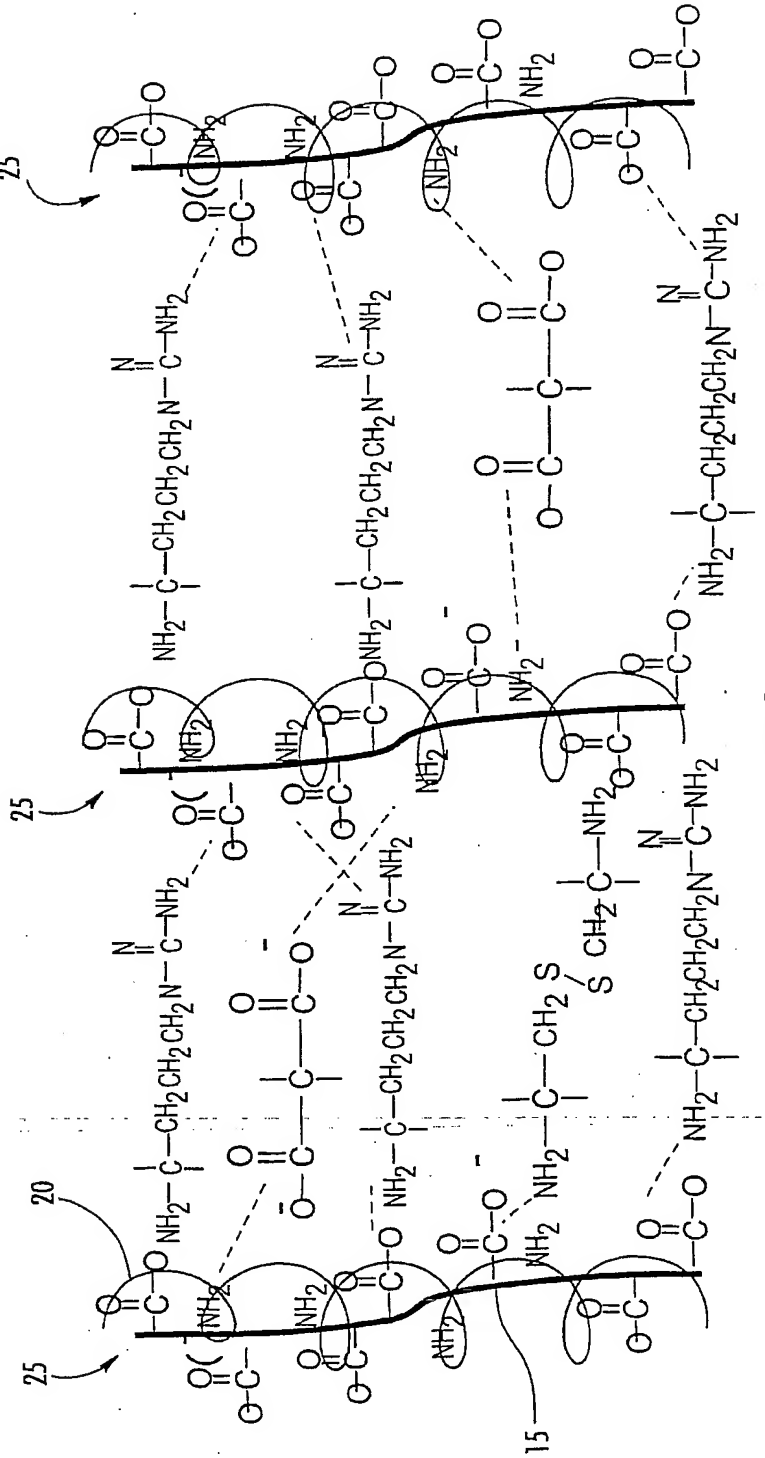


FIGURE 3

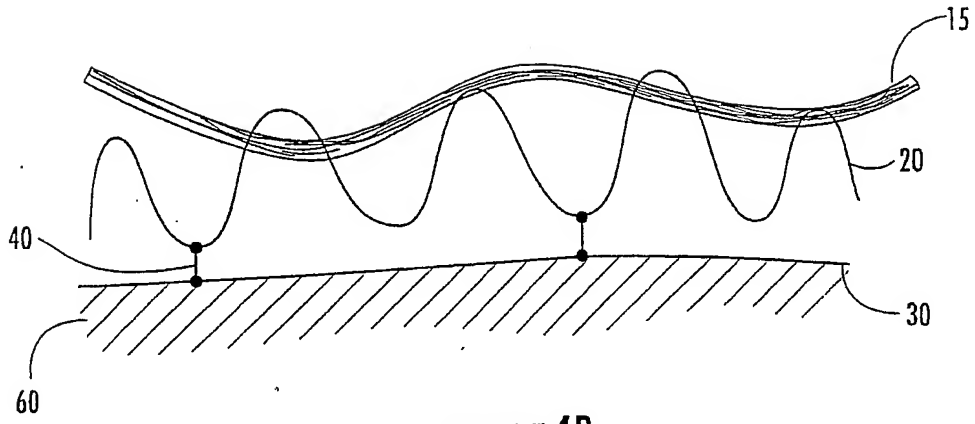


FIGURE 4B

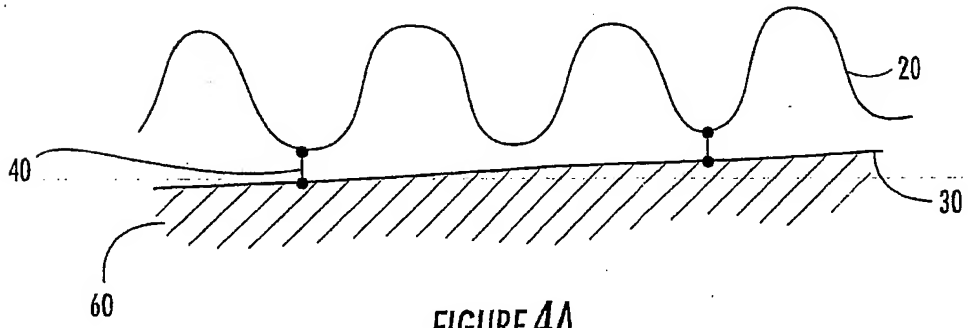


FIGURE 4A

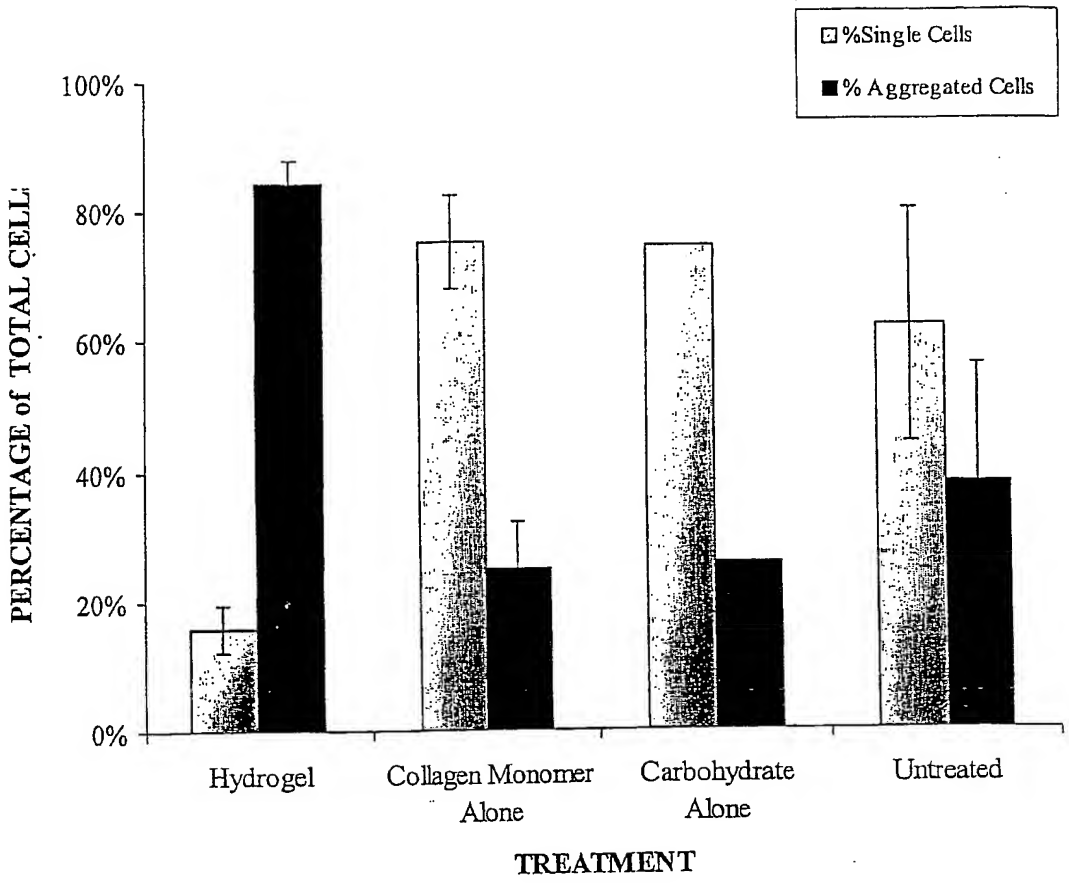

%Single Cells
% Aggregated Cells
100%
80%
60%
40%
20%
0%
PERCENTAGE of TOTAL CELL
Hydrogel
Collagen Monomer Alone
Carbohydrate Alone
Untreated
TREATMENT

FIGURE 5

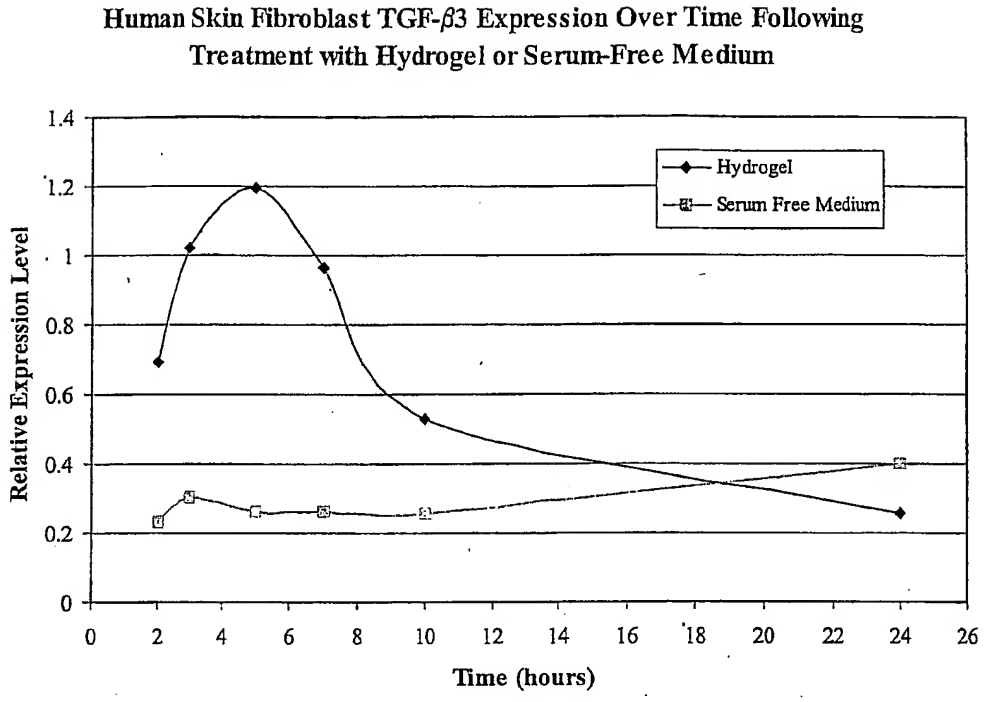

Human Skin Fibroblast TGF-β3 Expression Over Time Following Treatment with Hydrogel or Serum-Free Medium
Relative Expression Level
1.4
1.2
1
0.8
0.6
0.4
0.2
0
0 2 4 6 8 10 12 14 16 18 20 22 24 26
Time (hours)
Hydrogel
Serum Free Medium

FIGURE 6

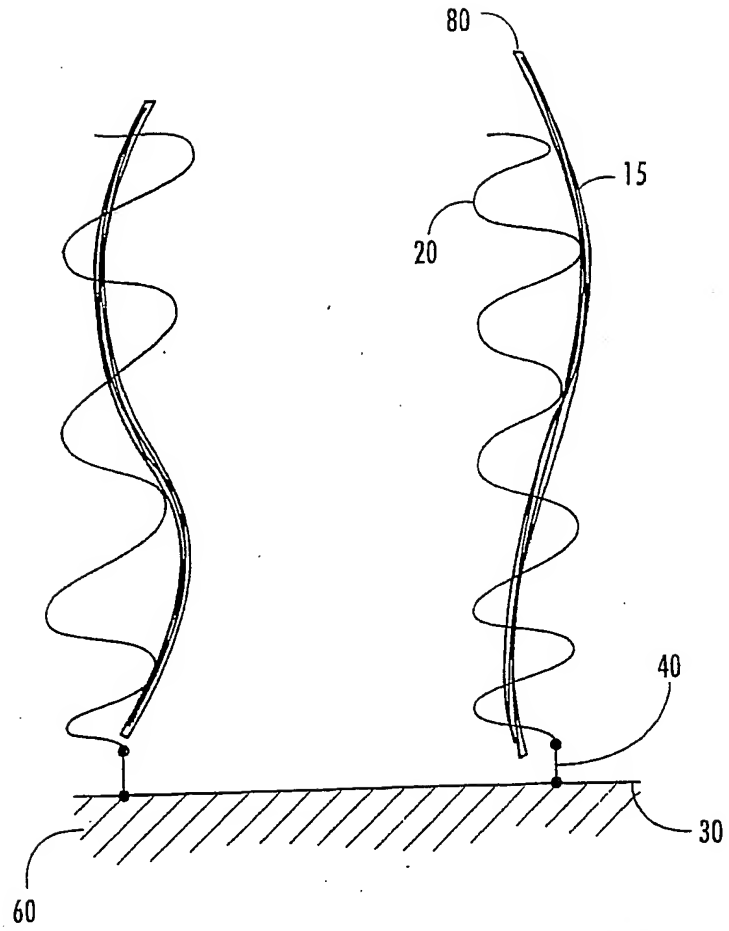


FIGURE 7

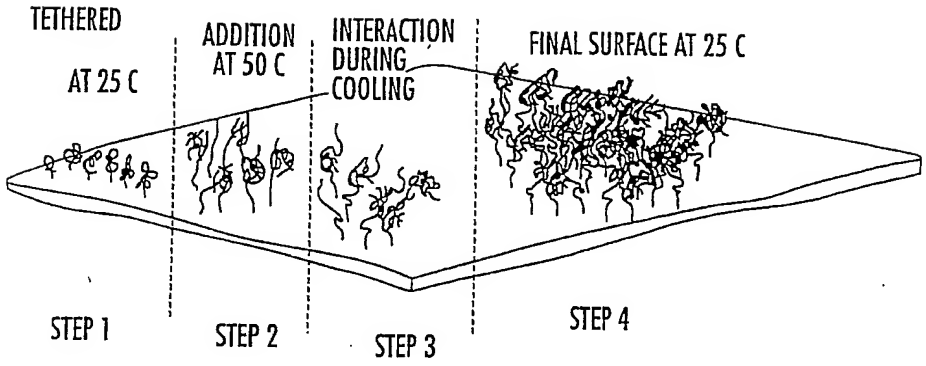

TETHERED
AT 25 C
ADDITION
AT 50 C
INTERACTION
DURING
COOLING
FINAL SURFACE AT 25 C
STEP 1
STEP 2
STEP 3
STEP 4

FIGURE 8

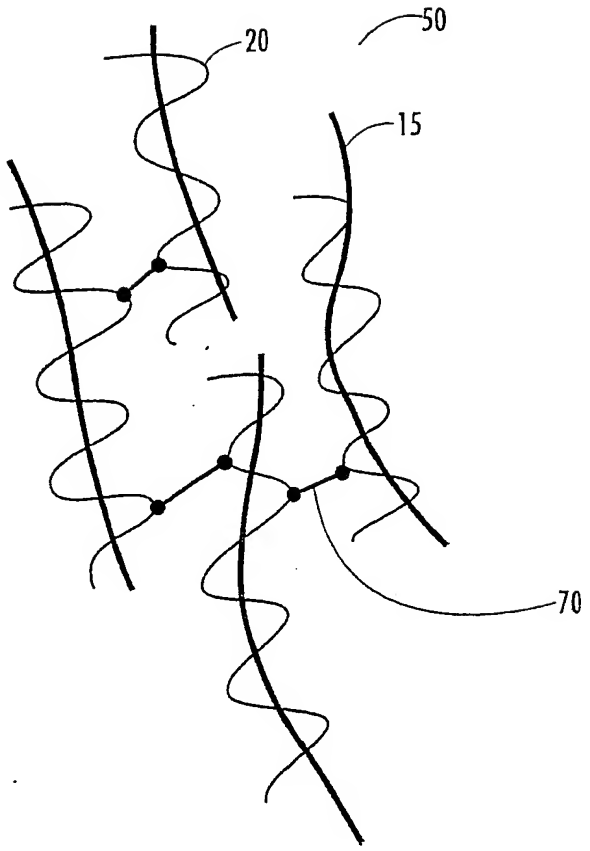


FIGURE 9